RADIOLOGICAL IMAGING APPARATUS

BACKGROUND OF THE INVENTION

5   The present invention relates to a radiological imaging apparatus, and more particularly, to a radiological imaging apparatus ideally applicable to radiological two-dimensional image pickup apparatus, X-ray computed tomography (hereinafter referred to as

10 "X-ray CT"), positron emission computed tomography (hereinafter referred to as "PET") and single photon emission computed tomography (hereinafter referred to as "SPECT").

        Among typical radiological imaging

15 apparatuses, which is a non-invasive imaging technology for examining functions and conformation of the body of a medical examinee, are radiological two-dimensional image pickup apparatus, X-ray CT, PET and SPECT, etc.

        PET inspection is an inspection consisting of

20 administering radiopharmaceutical (hereinafter referred to as "PET radiopharmaceutical") including positron emitters (^{15}O, ^{13}N, ^{11}C, ^{18}F, etc.) to the examinee and examining locations in the body where more PET radiopharmaceutical is consumed. The PET inspection is

25 an action of detecting γ-rays emitted from the body of the examinee caused by PET radiopharmaceutical using a radiation detector. More specifically, one positron emitted from a positron emitter in the PET

radiopharmaceutical couples with an electron of a
neighboring cell (cancerous cell), disappears and at
the same time irradiates a pair of γ-rays (called "γ-ray
pair") having energy of 511 keV. These γ-rays are
emitted in directions opposite to each other (180°
±0.6°). Detecting this pair of γ-rays using a radiation
detector makes it possible to know between which pair
of radiation detectors the positron is emitted.
Detecting those many γ-ray pairs makes it possible to
identify locations where more PET radiopharmaceutical
is consumed. For example, when PET radiopharmaceutical
created by coupling positron emitters and carbohydrate
is used, it is possible to discover cancer focuses
having hyperactive carbohydrate metabolism. One
example of the radiological imaging apparatus used for
PET is described in JP-A-7-20245. The data obtained is
converted to data of each voxel using the Filtered Back
Projection method described in non-patent document 1
(IEEE Transaction on Nuclear Science, Vol. NS-21, pp.
228-229). Positron emitters (^{15}O, ^{13}N, ^{11}C and ^{18}F, etc.)
used for the PET inspection have a half life of 2 to
110 minutes.

The SPECT administers radiopharmaceutical
(hereinafter referred to as "SPECT
radiopharmaceutical") including single photon emitters
(^{99}Tc, ^{67}Ga, ^{201}Tl, etc.) and matters (e.g., carbohydrate)
having a property of concentrating on a specific tumor
or specific molecule to an examinee and detects γ-rays

emitted from the emitters using a radiation detector. The energy of γ-rays emitted from the single photon emitters often used for inspection using the SPECT is around several 100 keV. In the case of the SPECT,

5 single γ-rays are emitted, and therefore it is not possible to obtain their angle of incidence upon the radiation detector. Thus, angle information is obtained by detecting only γ-rays incident from a specific angle through a collimator using the radiation

10 detector. The SPECT is an inspection method of identifying locations where more SPECT radiopharmaceutical is consumed by detecting γ-rays generated in the body caused by the SPECT radiopharmaceutical. One example of the radiological

15 imaging apparatus used for the SPECT is described in JP-A-9-5441. The SPECT also converts data obtained to data of each voxel using a method such as Filtered Back Projection. The SPECT may also take transmission images. ^{99}Tc, ^{67}Ga and ^{201}Tl used for the SPECT have a

20 half life longer than that of radionuclide used for the PET, for example, 6 hours to 3 days.

SUMMARY OF THE INVENTION

There is a demand for further improvement of diagnostic accuracy in the position and size, etc., of

25 an affected area such as malignant tumor and there is also a demand for improvement of accuracy of images including the affected area created by a radiological

imaging apparatus. It is also an important challenge to allow a damaged radiation detector to be replaced in a short time.

It is an object of the present invention to provide a radiological imaging apparatus capable of improving the accuracy of an image created and easily replacing a damaged radiation detector.

A feature of the present invention to attain the above-described object is that a detection unit is attached to a detector support member in a detachable manner and the detection unit is provided with a plurality of radiation detectors that detect radiation and other radiation detectors that detect radiation which has passed through some of the above described radiation detectors.

Since the other radiation detectors for detecting radiation that has passed through some radiation detectors are provided, it is possible to detect radiation emitted from the examinee by the some radiation detectors or the other radiation detectors and accurately confirm the position the radiation has reached from the some radiation detectors facing the examinee in the depth direction (position at which the radiation has been detected). Thus, it is possible to obtain an accurate image illustrating the condition of the body of the examinee. Furthermore, since the detector unit is attached to the detector support member in a detachable manner, a damaged radiation

detector can be replaced easily.

Other objects, features and advantages of the invention will become apparent from the following description of the embodiments of the invention taken
5   in conjunction with the accompanying drawings.

BRIEF DESCRIPTION OF THE DRAWINGS

Fig. 1 is a longitudinal sectional view of a radiological imaging apparatus of Embodiment 1 which is a preferred embodiment of the present invention;
10   Fig. 2 is a sectional view along a line II-II of Fig. 1;

Fig. 3 is an enlarged view of the section III of Fig. 1;

Fig. 4 is a sectional view along a line IV-IV
15 of Fig. 3;

Fig. 5 is a perspective view of the detector unit in Fig. 1;

Fig. 6 is a sectional view of the detector support member in the radius direction of the detector
20 unit in Fig. 5;

Fig. 7 shows a detailed structure of the signal discriminator in Fig. 1;

Fig. 8 illustrates a procedure for creating a tomographic image executed by the computer in Fig. 1;
25   Figs. 9A and 9B illustrate a state of γ-ray detection in the embodiment shown in Fig. 1;

Fig. 10 is a perspective view of a detector

unit applied to a radiological imaging apparatus according to Embodiment 2 which is another embodiment of the present invention;

Fig. 11 is a sectional view of the detector support member in the radius direction of the detector unit in Fig. 10;

Fig. 12 is a perspective view of a detector unit applied to a radiological imaging apparatus according to Embodiment 3 which is a further embodiment of the present invention; and

Fig. 13 is a sectional view of the detector support member in the radius direction of the detector unit in Fig. 11.

DESCRIPTION OF THE INVENTION

(Embodiment 1)

With reference now to Fig. 1 and Fig. 2, a radiological imaging apparatus which is a preferred embodiment of the present invention will be explained below. A radiological imaging apparatus 1 of this embodiment is used for a PET inspection. The radiological imaging apparatus 1 is provided with an image pickup apparatus 2, a signal processing apparatus 40, a tomographic image creation apparatus 35, an examinee holding apparatus 18, a drive apparatus control apparatus 21 and an X-ray source control apparatus 22. The examinee holding apparatus 18 includes a bed 20 on top of a bed support section 19 in

such a way that the bed 20 is movable in the longitudinal direction.

The image pickup apparatus 2 includes a casing 3, many detector units 4, a ring-shaped detector support member 8 and an X-ray source circumferential transport apparatus 13. As shown in Fig. 3 and Fig. 4, the detector support member 8 includes a ring-shaped detector support section 23 attached to a support member 39 and a cover member 24. The cover member 24 is attached to the detector support section 23, covering a signal discrimination unit housing space 44 formed in the detector support section 23.

The X-ray source circumferential transport apparatus 13 is provided with a guide rail 12 and an X-ray source apparatus 14. The ring-shaped guide rail 12 is attached to the examinee holding apparatus 18 on the side of the detector support member 8, more specifically, on the side of the detector support section 23 so as to surround a through hole section 41 into which the bed 20 is inserted. The X-ray source apparatus 14 includes an X-ray source drive apparatus 15, a telescopic arm 16 and an X-ray source 17. The X-ray source drive apparatus 15 is attached to the guide rail 12 in a movable manner. The X-ray source drive apparatus 15 includes a pinion (not shown) which engages with the rack of the guide rail 12 and a motor which rotates this pinion through a deceleration mechanism. The telescopic arm 16 is attached to a

casing (not shown) of the X-ray source drive apparatus 15 and can zoom in and out in the horizontal direction. The X-ray source 17 is attached to the end of the telescopic arm 16.

5   Though not shown, the X-ray source 17 includes a publicly known X-ray tube. This X-ray tube is provided with an anode, a cathode, a current source for the cathode and a voltage source for applying a voltage to between the anode and cathode inside the

10 external cylinder. The cathode is a tungsten filament. When a current flows from the current source to the cathode, electrons are emitted from the filament. These electrons are accelerated by a voltage (several hundred kV) applied from the voltage source to between

15 the cathode and anode and collide with the anode (W, Mo, etc.), the target. Collision of electrons with the anode produces X-rays of 80 keV. These X-rays are radiated from the X-ray source 17.

          The tomographic image creation apparatus 35

20 is provided with a computer 36 and a storage apparatus 37. The computer 36 is connected to a coincidence counter 34 and the storage apparatus 37 is connected to the computer 36. The computer 36 is a tomographic image creation section. A display device 38 is

25 connected to the computer 36.

          As shown in Fig. 5 and Fig. 6, the detector unit 4 has a structure in which a plurality (e.g., 9) of radiation detectors 5 are arranged on one side of

the support substrate 6 and a connector section 7 is attached to the support substrate 6. Nine radiation detectors 5 are arranged in 3 rows and 3 columns on the support substrate 6. The "circumferential direction"

5   shown in Fig. 4 refers to the circumferential direction of the detector support member 8, the "axial direction" refers to the axial direction of the detector support member 8 and the "radius direction" refers to the "radius direction of the detector support member 8 (the

10   same applies to Fig. 10 and Fig. 12). Cathode electrodes K1, K2 and K3 of one column of the three radiation detectors 5 arranged in the radius direction of the detector support member 8, that is, radiation detectors 5A, 5B and 5C are connected to a grounding

15   wire 45. The grounding wire 45 is connected to a connector terminal 7D of the connector section 7. A wire 46 connected to an anode electrode A1 of the radiation detector 5A is connected to a connector terminal 7A. A wire 47 connected to an anode electrode

20   A2 of the radiation detector 5B is connected to a connector terminal 7B of the connector section 7. Furthermore, a wire 48 connected to an anode electrode A3 of the radiation detector 5C is connected to a connector terminal 7C of the connector section 7. The

25   radiation detectors 5 included in other two columns are likewise connected to the other connector terminals provided for the connector section 7. The grounding wire 45 and wires 46, 47 and 48 are all set in the

support substrate 6. The many detector units 4 are set
and held in the detector support section 23 by engaging
their respective connector terminals such as the
connector terminal 7A with the connector section 11
5   provided in the detector support section 23. The
detector units 4 surround the through hole section 41
and many of them are arranged in the circumferential
direction and axial direction of the through hole
section 41. These detector units 4 are attached to the
10 detector support section 23 in a detachable manner.

The casing 3 is attached to the detector
support section 23 so as to cover the detector units 4
(Fig. 4). Furthermore, the casing 3 extends in the
horizontal direction and forms the hole section
15 (through hole section) 41 into which the bed 20 is
inserted during an inspection (Fig. 1).

With the arrangement of these detector units
4, many radiation detectors (e.g., 10,000 in total) 5
are arranged in the casing 3 on the inner side of the
20 ring-shaped detector support member 8. These radiation
detectors 5 are arranged in the radius direction of the
detector support member 8 in multilayers (e.g., three
layers) and also arranged in the axial direction of the
detector support member 8 over a plurality of columns.
25 The three radiation detectors 5 (radiation detectors
5A) located farthest from the connector section 7 of
the radiation detectors 5 arranged in the respective
detector units 4 are located closest to the central

axis of the through hole section 41 and are called "first layer radiation detectors." The three radiation detectors 5 (radiation detectors 5C) located closest to the connector section 7 are located farthest from the

5 central axis of the through hole section 41 and are called "third layer radiation detectors." The three radiation detectors 5 (radiation detectors 5B) located between the first layer and third layer in the detector unit 4 are called "second layer radiation detectors."

10 The radiation detection apparatus 43 includes the aforementioned many radiation detection units 4. The radiation detection apparatus 43 includes many radiation detectors 5 arranged in the radius direction of the detector support member 8 from the first layer

15 to the third layer and arranged in the axial direction of the detector support member 8.

Examples of typical radiation detectors include a semiconductor radiation detector and scintillator. For the scintillator, a photoelectron

20 multiplier, etc., needs to be arranged behind a crystal (BGO, NaI, etc.) which is a radiation detection section, and therefore it is unsuitable for a multilayer arrangement (e.g., three layers as described above). Since the semiconductor radiation detector

25 requires no photoelectron multiplier, etc., it is suitable for a multilayer arrangement. In this embodiment, semiconductor radiation detectors are used for the radiation detectors 5 and a 5 mm cube which is

a detection section is made of cadmium telluride (CdTe). The detection section may also be made of gallium arsenide (GaAs) or cadmium zinc telluride (CZT).

The signal processing apparatus 40 includes a signal discriminator 27, γ-ray discriminator 32 and the coincidence counter 34. One signal discriminator 27 is provided for each radiation detector 5 on the first layer. Furthermore, one γ-ray discriminator 32 is provided for each radiation detector 5 on the second layer and the third layer respectively. These three signal discriminators 27 and six γ-ray discriminators 32 are set on one substrate 26. A signal discrimination unit 25 is made up of the three signal discriminators 27 and six γ-ray discriminators 32 set on one substrate 26. The substrate 26 is attached to a unit support member 66. Each signal discrimination unit 25 provided for every detector unit 4 is attached to the unit support member 66 arranged in a signal discrimination unit housing space 44 as shown in Fig. 4. The unit support member 66 is attached to the detector support member 23. The signal discrimination unit 25 is set in the unit support member 66 and is thereby supported to the detector support member 8. It is also possible to attach the substrate 26 directly to the detector support member 23 as the support substrate without using the unit support member 66.

As shown in Fig. 7, the signal discriminator

27 includes a changeover switch 28, a γ-ray

discriminator 32 and an X-ray signal processing

apparatus 33. The changeover switch 28 includes a

movable terminal 29 and fixed terminals 30 and 31. The

5 γ-ray discriminator 32 is connected to the fixed

terminal 30 and the X-ray signal processing apparatus

33 is connected to the fixed terminal 31. The

connector terminal 7A connected to the first layer

radiation detector 5A has contact with a connector

10 terminal 11A provided for the connector section 11

through the engagement of the connector section 7 with

the connector section 11. The movable terminal 29 is

connected to the connector terminal 11A through a wire

49. The wire 49 is set in the unit support member 66.

15 The minus terminal of a power supply 50 is connected to

the wire 46 and the plus terminal of the power supply

50 is connected to the radiation detector 5A. The γ-ray

discriminators 32 in all the signal discriminators 27

are connected to the coincidence counter 34 through a

20 wire 52. On the other hand, the X-ray signal

processing apparatuses 33 in all the signal

discriminators 27 are connected to the computer 36

through a wire 53.

Three of the six γ-ray discriminators 32 other

25 than the signal discriminator 27 provided in the signal

discrimination unit 25 are connected to a connector

terminal 11B of the connector section 11 (not shown)

through a wire 54. The connector terminal 11B contacts

the connector terminal 7B to which the second layer

radiation detectors 5B are connected. The remaining

three γ-ray discriminators 32 are connected to a

connector terminal 11C (not shown) of the connector

5   section 11 through another wire 54. The connector

terminal 11B contacts the connector terminal 7C to

which the third layer radiation detectors 5C are

connected. The six γ-ray discriminators 32 other than

the signal discriminator 27 are each connected to the

10 coincidence counter 34 through a wire 55. Fig. 1 shows

the signal discrimination unit 25 and wires 54 and 56

outside the detector support member 8 and this is

intended to make the wiring connection states of the

signal discriminator 27 and γ-ray discriminators 32

15 provided for the signal discrimination unit 25 easy to

understand. The signal discrimination unit 25 is

actually set in the detector support member 8 as shown

in Fig. 3 and Fig. 4 and the wires 52, 53 and 55 are

drawn out of the detector support member 8.

20   Before specifically explaining the

radiological inspection in this embodiment, the

principle of radiation detection of this embodiment

will be explained. Data of an X-ray CT image

(tomographic image including an image of internal

25 organs and bones of an examinee obtained through X-ray

CT) is created by irradiating X-rays radiated from the

X-ray source onto an examinee in a specific direction

for a predetermined time and repeating an operation of

detecting (scanning) X-rays that have passed through the body using radiation detectors and based on the intensity of X-rays detected by a plurality of radiation detectors. In order to obtain accurate X-ray

5　CT image data, it is preferable not to allow γ-rays emitted from the inside of the examinee caused by PET radiopharmaceutical to enter the radiation detectors which are detecting X-rays during an X-ray CT inspection. With regard to one radiation detector, if

10　the time of X-ray irradiation onto the examinee is shortened in accordance with a rate of incidence of γ-rays, influences of γ-rays can be ignored, and therefore this system is designed to shorten the time of X-ray irradiation onto the examinee. To calculate the X-ray

15　irradiation time T, a rate of incidence of γ-rays on one radiation detector is considered first. Suppose radioactivity in the body based on PET radiopharmaceutical administered to the examinee in a PET inspection is N (Bq), rate of generated γ-rays

20　passing through the body is A, the rate of incidence calculated from the solid angle of one radiation detector is B and sensitivity of the detection element is C. Then, rate α (counts/sec) of γ-rays detected by one radiation detector is given by Expression (1).

25　　　　　$\alpha = 2NABC$　　... (1)

In Expression (1), coefficient "2" means that a pair (two) of γ-rays are emitted for annihilation of one positron. The probability W that γ-rays are detected by

one detection element within the irradiation time T is given by Expression (2).

$$W = 1-\exp(-T\alpha) \ldots (2)$$

By determining the irradiation time T so that the value of W in Expression (2) is reduced, the influence of γ-rays incident on one radiation detector is reduced to a level as small as negligible during an X-ray CT inspection.

An example of the X-ray irradiation time T will be explained below. A specific X-ray irradiation time T was calculated based on Expressions (1) and (2). The intensity of radiation in the body caused by PET radiopharmaceutical administered to the examinee in a PET inspection is a maximum of approximately 370 MBq (N = 370 MBq) and the rate of passage A of γ-rays through the body is on the order of 0.6 (A = 0.6) when the body of the examinee is assumed to be water having a radius of 15 cm. For example, assuming a case where a radiation detector having a size of 5 mm per side is arranged in the shape of a ring having a radius of 50 cm, the rate of incidence B calculated from the solid angle of one radiation detector is 8×10^{-8} (B = 8×10^{-8}). Furthermore, detection sensitivity C of the radiation detector is a maximum of approximately 0.6 (C = 0.6) when a semiconductor radiation detector is used. From these values, the γ-ray detection rate α of one radiation detector is on the order of 2000 (counts/sec). Suppose the X-ray irradiation time T is

1.5 μsec, for example. Then, the probability W that one radiation detector will detect γ-rays during X-ray detection is 0.003 and most of these γ-rays can be ignored. When radioactivity in the body is 370 MBq or

5 less, if the X-ray irradiation time is 1.5 μsec or less, W < 0.003, that is, the probability W of γ-ray detection is 0.3% or less, which is negligible.

An X-ray CT inspection and a PET inspection in this embodiment where the above described principle

10 is applied and the image pickup apparatus 2B is used will be explained more specifically.

An X-ray CT inspection and a PET inspection in this embodiment will be explained. PET radiopharmaceutical is administered to the examinee 42

15 using a method like injection in such a way that radioactivity in the body becomes 370 MBq. Then, the examinee 42 waits for a predetermined time until the PET radiopharmaceutical is spread in the body of the examinee 42, concentrated on an affected area (e.g.,

20 affected area of cancer) and image pickup is possible. PET radiopharmaceutical is selected according to the affected area to be inspected. After the lapse of the predetermined time, the bed 20 on which the examinee 42 lies is inserted into the through hole section 41 of

25 the image pickup apparatus 2 together with the examinee 42. An X-ray CT inspection and a PET inspection are carried out using the image pickup apparatus 2. The examinee 42 with the PET radiopharmaceutical

administered is inserted into the through hole section
41, a voltage is applied from the power supply 50 to
the respective radiation detectors 5 and then the
radiation detectors 5 detect γ-rays emitted from the
5 examinee 42. That is, a PET inspection is started.
After the PET inspection is started, an X-ray CT
inspection is started.

The X-ray CT inspection will be explained.
When the X-ray CT inspection is started, the drive
10 apparatus control apparatus 21 outputs a drive start
signal and closes the breaker (hereinafter referred to
as "motor breaker") connected to the motor of the X-ray
source drive apparatus 15 and connected to the power
supply. The torque of the motor is transmitted to the
15 pinion through the deceleration mechanism and the X-ray
source apparatus 14, that is, the X-ray source 17 moves
along the guide rail 12 in the circumferential
direction. The X-ray source 17 moves around the
examinee 42 at a set speed while being inserted in the
20 through hole section 41. When the X-ray CT inspection
is completed, the drive apparatus control apparatus 21
outputs a drive stop signal and opens the motor
breaker. This stops the movement of the X-ray source
17 in the circumferential direction. In this
25 embodiment, all the radiation detectors 5 move neither
in the circumferential direction nor in the axial
direction of the through hole section 41. The drive
apparatus control apparatus 21 and X-ray source control

apparatus 22 are set in the detector support member 8.
A publicly known technology is applied to transmission
of a control signal from the drive apparatus control
apparatus 21 and X-ray source control apparatus 22 to

5   the moving X-ray source apparatus 14 without
interfering with the movement of the X-ray source
apparatus 14.

The X-ray source control apparatus 22
controls the time of radiation of X-rays from the X-ray

10 source 17. That is, the X-ray source control apparatus
22 outputs X-ray generation signals and X-ray stop
signals repeatedly. The first X-ray generation signal
is outputted based on the input of the above drive
start signal to the X-ray source control apparatus 22.

15 In response to the output of an X-ray generation
signal, the breaker provided between the anode (or
cathode) of the X-ray tube of the X-ray source 17 and
the power supply (hereinafter referred to as "X-ray
source breaker", not shown) is closed, an X-ray stop

20 signal is outputted after a lapse of a first set time,
the X-ray source breaker is opened and the X-ray source
breaker is closed after a lapse of a second set time,
and this control is repeated. Between the anode and
cathode, a voltage is applied during the first set

25 time, but no voltage is applied during the second set
time. With such control by the X-ray source control
apparatus 22, pulse-shaped 80 keV X-rays are radiated
from the X-ray tube. An irradiation time T which is

the first set time is set, for example, to 1 μsec so that the probability of detection of γ-rays at the radiation detector 5 can be ignored. The second set time is a time T0 during which the X-ray source 17

5   moves between one radiation detector 5 and another radiation detector 5 which is adjacent thereto in the circumferential direction and is determined by the moving speed of the X-ray source 17 in the circumferential direction of the guide rail 12. The

10 first and second set times are stored in the X-ray source control apparatus 22.

In response to repeated outputs of X-ray stop signals and X-ray generation signals, the X-ray source 17 radiates X-rays during the first set time, that is,

15 for 1 μsec and stops radiation of X-rays during the second set time. This radiation and stop of X-rays are repeated during a period during which the X-ray source 17 is moving in the circumferential direction.

The X-rays 57 radiated from the X-ray source

20 17 are irradiated onto the examinee 42 in a fan-beam shape. The examinee 42 receives irradiation of X-rays 57 as the X-ray source 17 moves in the circumferential direction. The X-rays 57 which have passed through the examinee 42 (e.g., X-rays that have passed through the

25 affected area 56) are detected by a plurality of radiation detectors 5 located in the circumferential direction centered on the radiation detector 5 located at the position 180° from the X-ray source 17 relative

to the central axis of the through hole section 41.
These radiation detectors 5 output the detection
signals of the X-rays 57. These X-ray detection
signals are inputted to their respective signal

5   discriminators 27 through the corresponding wires 49.
The radiation detectors 5 that detect the above
described X-rays are referred to as first radiation
detectors 5 for the purpose of convenience.

From the affected area (affected area of

10   cancer) of the examinee 42 on the bed 16, γ-rays 58 of
511 keV caused by PET radiopharmaceutical are emitted.
The radiation detectors 5 other than the first
radiation detectors 5 detect the γ-rays 58 and output γ-
ray detection signals. The radiation detectors 5

15   detecting γ-rays are called "second radiation detectors
5" for the purpose of convenience. The γ-ray detection
signals outputted from the second radiation detectors 5
located on the first layer out of the second radiation
detectors are inputted to their respective signal

20   discriminators 27 through the corresponding wires 49
and the γ-ray detection signals outputted from the
second radiation detectors 5 located on the second
layer and third layer are inputted to their respective
γ-ray discriminators 32 through the corresponding wires

25   54. Only the radiation detectors 5 located on the
first layer are connected to the signal discriminator
61 having the X-ray signal processing apparatus 33.
This is because energy of the X-rays is 80 keV and most

(90% or more) of the X-rays that have passed through the examinee 42 are detected by the first layer radiation detectors 5.

In the signal discriminators 27, the γ-ray detection signals outputted from the first layer second radiation detectors 5 are transmitted to the γ-ray discriminators 32 and the X-ray detection signals outputted from the first radiation detectors 5 are transmitted to the X-ray signal processing apparatus 33. Such transmission of each detection signal is performed according to the changeover operation of the changeover switch 28 of the signal discriminator 27. The changeover operation for connecting the movable terminal 29 of the changeover switch 28 to the fixed terminal 30 or fixed terminal 31 is performed based on the changeover control signal which is the output of the drive apparatus control apparatus 21. During an X-ray CT inspection, the drive apparatus control apparatus 22 selects the first radiation detectors 5 of the first layer radiation detectors 5 and connects the movable terminal 29 of the signal discriminators 27 connected to these first radiation detector 5 to the fixed terminal 31.

The selection of the first radiation detectors 5 will be explained. The motor in the X-ray source drive apparatus 15 is connected with an encoder (not shown). The drive apparatus control apparatus 22 receives a detection signal of the encoder and

calculates the position of the X-ray source drive

apparatus 15 in the circumferential direction of the

detector support member 8 (through hole section 41),

that is, the position of the X-ray source 17 and

5   selects the radiation detector 5 located 180° opposite

to the position of the X-ray source 17 using the stored

data of the position of each radiation detector 5. The

X-rays 57 radiated from the X-ray source 17 has a

certain width in the circumferential direction of the

10   guide rail 12, and therefore there are a plurality of

radiation detectors 5 in the circumferential direction

other than the selected radiation detectors 5 which

detect the X-rays 57 that have passed through the

examinee 42. The drive apparatus control apparatus 22

15   selects those plurality of radiation detectors 5, too.

These radiation detectors 5 are the first radiation

detectors 5. The first radiation detectors 5 also

change depending on the movement of the X-ray source 17

in the circumferential direction. As the X-ray source

20   17 moves in the circumferential direction, the first

radiation detectors 5 also seem to move in the

circumferential direction. When the drive apparatus

control apparatus 22 selects other radiation detectors

5 as the X-ray source 17 moves in the circumferential

25   direction, the movable terminal 29 connected to the

radiation detectors 5 which become the new first

radiation detectors 5 is connected to the fixed

terminal 31. The movable terminal 29 connected to the

radiation detectors 5 which are no longer the first

radiation detectors 5 as the X-ray source 17 moves in

the circumferential direction is connected to the fixed

terminal 30 by the drive apparatus control apparatus

5   22.   According to the positional relationship with the

X-ray source 17, the first layer radiation detectors 5

sometimes become the first radiation detectors 5 and

sometimes become the second radiation detectors 5.   For

this reason, one radiation detector 5 on the first

10   layer outputs both an X-ray detection signal and γ-ray

detection signal with a time shift in between.

        The first radiation detectors 5 receive

irradiation from the X-ray source 17 for 1 μsec which

is the first set time and detect X-rays which have

15   passed through the examinee 42.   The probability that

the first radiation detectors 5 detect γ-rays emitted

from the examinee 42 for 1 μsec is as small as

negligible as described above.   Many γ-rays 58 generated

at the affected area 56 of the examinee 42 caused by

20   PET radiopharmaceutical are not emitted in a specific

direction but emitted in all directions.   These γ-rays

58 form pairs and are emitted in directions opposite to

each other (180° ±0.6°) as described above and detected

by either one of the second radiation detectors 5.

25   The signal processing of the signal

discriminators 27 when the X-ray detection signals and

γ-ray detection signals outputted from the first layer

radiation detectors 5 are inputted will be explained.

As described above, the X-ray detection signals
outputted from the first layer radiation detectors 5
are inputted to the X-ray signal processing apparatus
33. The X-ray signal processing apparatus 33

5   integrates the input X-ray detection signals using an
integration apparatus and outputs the X-ray detection
signal integrated value, that is, information on the
intensity of the measured X-rays. The intensity
information of the X-ray detection signals is

10   transmitted to the computer 26 through the wire 53 and
stored in the storage apparatus 37.

The γ-ray detection signals outputted from the
first layer second radiation detectors 5 are inputted
to the γ-ray discriminators 32 by an action of the

15   changeover switch 28. By annihilation of positrons
emitted from PET radiopharmaceutical, energy of γ-rays
emitted from the affected area 56 is 511 keV. However,
when γ-rays are scattered in the body of the examinee
42, the energy falls below 511 keV. To remove

20   scattered γ-rays, the γ-ray discriminators 32 is
provided with a filter (not shown) which allows γ-ray
detection signals having energy equal to or greater
than an energy set value of 400 keV which is lower than
511 keV to pass. This filter receives the γ-ray

25   detection signals outputted from the fixed terminal 30.
Here, 400 keV was used as the energy set value because
variations of the γ-ray detection signals produces when
γ-rays of 511 keV enter the radiation detectors 5 are

taken into consideration. The γ-ray discriminators 32 generate a pulse signal having predetermined energy when γ-ray detection signals having energy equal to or greater than the energy set value (400 keV) are

5 inputted. The γ-ray discriminator 32 is a γ-ray detection signal processing apparatus and gives the pulse signal to be output time information and position information indicating the position of the radiation detector 5 connected to the γ-ray discriminator 32. The

10 time information is either the information when γ-ray detection signals are inputted to the γ-ray discriminator 32 or information when a pulse signal is outputted from the γ-ray discriminator 32.

All the second layer and third layer

15 radiation detectors 5 are the second radiation detectors. The γ-ray discriminators 32 connected to these second layer and third layer radiation detectors 5 through the wire 54 also exhibits the same functions as those of the γ-ray discriminators 32 in the above

20 described signal discriminators 27.

The coincidence counter 34 receives pulse signals outputted from all the γ-ray discriminators 32. The coincidence counter 34 performs simultaneous counting using pulse signals for respective γ-ray

25 detection signals outputted from the two second radiation detectors 5 which have detected each γ-ray 58 of the γ-ray pair (a pair of the second radiation detectors which exist at substantially 180° (more

precisely 180° ±0.6°) different positions centered on the central axis of the through hole section 41) and calculates the count rate (γ-ray count information) corresponding to their respective γ-ray detection

5 signals. The coincidence counter 34 decides whether each pulse signal corresponds to the respective γ-ray detection signals of the γ-ray pair or not based on the time information given to each pulse signal. That is, if the difference between two time information pieces

10 is within a set time (e.g., 10 nsec), the coincidence counter 34 decides that the pulse signal corresponds to a pair of γ-rays 58 generated by annihilation of one positron. Furthermore, the coincidence counter 34 creates data from the position information given to

15 those pulse signals as the position of the corresponding pair of second radiation detectors 5, that is, the position information of each γ-ray detection point. The coincidence counter 34 outputs the count rate information corresponding to each γ-ray

20 detection signal and the position information of the two detection points at which the γ-ray pair is detected. The count rate and position information are transmitted to the computer 36 and stored in the storage apparatus 37.

25 The computer 36 carries out processing based on the processing procedure in steps 60 to 65 shown in Fig. 8. The computer 36 that carries out such processing is a tomography creation section that

creates first tomographic image information using first
information (more specifically, γ-ray count information
and position information of the γ-ray detection point),
creates second tomographic image information (more

5 specifically, X-ray CT image data) using second
information (more specifically, X-ray intensity
information and X-ray detection position information)
and third tomographic image information (more
specifically, combined tomography data) including the

10 first tomographic image information and second
tomographic image information using those tomographic
image information pieces. The count rate information
of the γ-ray detection signal counted by the coincidence
counter 34, position information of the γ-ray detection

15 signal outputted from the coincidence counter 34, X-ray
intensity information outputted from the X-ray signal
processing apparatus 33 and X-ray detection position
information given to the X-ray intensity are inputted
(step 60). The inputted count rate information of the

20 γ-ray detection signal, position information of the γ-
ray detection point, X-ray intensity information and X-
ray detection position information are stored in the
storage apparatus 37 (step 61).

The tomographic image of the cross section

25 (hereinafter the "cross section" will refer to the
cross section when the examinee is in standing posture)
of the examinee 42 is reconstructed using the X-ray
intensity information and X-ray detection position

information (step 62). The reconstructed tomographic image is called an "X-ray CT image." Specific processing of reconstruction of this tomographic image will be explained. First, the attenuation rate of X-rays in each voxel in the body of the examinee 42 is calculated using the X-ray intensity information. This attenuation rate is stored in the storage apparatus 37. To reconstruct the X-ray CT image, the linear attenuation coefficient in the body of the examinee 42 between the position of the X-ray source 17 and the position of the radiation detector 5 which has detected the X-rays (obtained from the X-ray detection position information) is calculated using the attenuation rate of the X-ray detection signal read from the storage apparatus 37. The position of the X-ray source 17 during movement detected by the encoder is given to the X-ray intensity information by each X-ray signal processing apparatus 33 and transmitted to the computer 36. The CT value of each voxel is calculated based on the value of the linear attenuation coefficient obtained by the filtered back projection method using the linear attenuation coefficient. The data of the X-ray CT image is obtained using those CT values and stored in the storage apparatus 37. In step 62, the X-ray CT image on the cross section that passes through the affected area where PET radiopharmaceutical is concentrated is also reconstructed.

The tomographic image of the cross section of

the examinee 42 including the affected area (e.g., the affected area of cancer) is reconstructed using the count rate of the γ-ray detection signal at the corresponding position (step 63). The tomographic

5   image reconstructed using the count rate of the γ-ray detection signal is called a "PET image". This processing will be explained in detail. Using the count rate of the γ-ray detection signal read from the storage device 37, the number of γ-ray pairs generated

10   (the number of γ-ray pairs generated according to annihilation of a plurality of positrons) in the body between the semiconductor devices of a pair of the second radiation detectors 5 (specified by position information of the γ-ray detection point) is calculated.

15   Using this number of γ-ray pairs generated, a γ-ray pair generation density of each voxel is calculated according to the filtered back projection method. PET image data can be obtained based on this γ-ray pair generation density. This PET image data is stored in

20   the storage device 37.

The PET image data is combined with the X-ray CT image data to obtain combined tomographic data including both data pieces and stored in the storage device 37 (step 64). The PET image data at the

25   position of the affected area and X-ray CT image data at the position are combined to obtain combined tomographic image data on the cross section of the examinee 42 at the position of the affected area. The

combination of the PET image data and X-ray CT image data can be performed easily and accurately by aligning the central axis of the through hole section 41 in both image data pieces. That is, the PET image data and X-

5 ray CT image data are created based on the detection signals outputted from the shared radiation detector 5, and therefore alignment can be performed accurately as described above. The combined tomographic data is called from the storage device 37 and output to the

10 display device 38 (step 65) and displayed on the display device 38. The combined tomographic image displayed on the display device 38 includes an X-ray CT image, and therefore it is possible to easily check the position in the body of the examinee 42 of the affected

15 area in the PET image. That is, since the X-ray CT image includes images of internal organs and bones, doctors can identify the position of the affected area (e.g., the affected area of cancer) from the relationship with the internal organs or bones.

20 The radiological imaging apparatus 1 comprises a plurality of radiation detectors 5 multilayered in the radius direction of the through hole section 41 (Fig. 1 to Fig. 4) and this multilayer arrangement can bring out the following new function.

25 For example, suppose a case where two γ-rays 58a and 58b emitted from a point of γ-ray pair generation 70 (in the affected area 56) in the body of the examinee 42 as shown in Fig. 9A enter radiation detectors 5f and 5g.

It is unknown at which position in the radiation detector the γ-rays have attenuated, and therefore the conventional method considers a line connecting the end positions of the pair of the radiation detectors 5f and

5   5h, that is, a line 71 shown in Fig. 9B as a detection line. However, since the radiological imaging apparatus 1 adopts the multilayer arrangement of the radiation detectors 5 in the radius direction of the through hole section 41, it is possible to obtain a γ-

10   ray detection signal of the radiation detector 5g located outside in the radius direction and consider a line 72 connecting the radiation detector 5f and radiation detector 5g as a detection line. That is, it is possible to grasp the attenuation position in the

15   depth direction of the radiation detectors 5 which have been unknown in the conventional example. As a result, the detection line 72 precisely passes through the position at which a pair of γ-rays are generated and therefore the accuracy of the image improves. As a

20   result, the detection line is brought closer to the actual point of generation of the pair of γ-rays and accuracy of measured data improves.

In this embodiment, the radiation detection apparatus 43 comprises a plurality of radiation

25   detectors 5 which output both X-ray detection signals and γ-ray detection signals, and therefore the radiation detection apparatus 43 functions as a γ-ray detection section as well as an X-ray detection section. That

is, the radiation detection apparatus 43 has both
functions of the γ-ray detection section and X-ray
detection section. In this embodiment, the X-ray
detection section is positioned in an area formed
5   between one end of the γ-ray detection section and the
other end of the γ-ray detection section in the
longitudinal direction of the bed 20. Furthermore, the
radiation detection apparatus 43 is an X-ray detection
section that detects X-ray 57 which is irradiated from
10   the X-ray source 17 and passes through the examinee 42
and outputs a detection signal of this X-ray 57 and at
the same time is a γ-ray detection section that detects
a γ-ray 58 emitted caused by PET radiopharmaceutical
from the area (affected area 56) through which the X-
15   ray 57 in the examinee 42 passes at the position of the
examinee 42 irradiated with the X-ray 57 and outputs a
detection signal of this γ-ray 58.

        This embodiment can attain the effects
described below.
20   (1)   In this embodiment, a plurality of detection
units 4 are attached to the detector support member 8
through the connector section, and therefore these
detection units 4, or more specifically, many radiation
detectors 5 can be attached in a short time. This
25   makes it possible to shorten the time for manufacturing
the image pickup apparatus 2, that is, the radiological
imaging apparatus 1.

        (2)   Since the detection units 4 are attached to

the detector support member 8 through the connector
section in a detachable manner, when a radiation
detector 5 has trouble, the detection unit 4 including
the radiation detector 5 in trouble can be easily

5   removed from the detector support member 8.
Furthermore, a new detection unit 4 can be easily
attached to the detector support member 8 at the
position of the removed detector unit. Thus, this
embodiment allows a damaged radiation detector 5 to be

10 replaced easily.

(3)   This embodiment arranges the plurality of
radiation detectors 5 not only in the axial direction
and circumferential direction of the through hole
section 41 (detector support member 8) but also in the

15 radius direction, and can thereby obtain γ-ray detection
signals at the positions subdivided in the radius
direction of the through hole section 41 without
reducing signal transmission substances as in the case
of conventional radiation detectors used for a PET

20 inspection. Thus, this embodiment can obtain precise
information on the position that γ-rays reach in the
radius direction of the through hole section 41
(position information of the radiation detectors 5
which have output γ-ray detection signals). In a

25 conventional PET inspection, one radiation detector is
placed in the radius direction of the through hole
section 41, a reflector is placed inside this radiation
detector and the information on the position that γ-rays

have reached in the radius direction of the through

hole section 41 is obtained according to a pattern with

which the signal transmission substance reaches the

photoelectron multiplier. At this moment, part of the

5   signal transmission substance attenuates in the

radiation detector because of the reflector and is

reflected to the outside of the radiation detector,

which reduces the signal transmission substance and

reduces energy resolution.

10   (4)   This embodiment arranges a plurality of

independent radiation detectors 5 in the radius

direction of the through hole section 41, and can

thereby use all the signal transmission substance of

the radiation detectors 5 for detection of γ-rays and

15   improve energy resolution of the radiation detectors 5.

Use of the radiation detectors 5 with high energy

resolution for a PET inspection makes it possible to

distinguish γ-rays whose energy has attenuated due to

scattering from un-scattered γ-rays having energy of 511

20   keV. As a result, it is possible to remove more

scattered radiation through a filter of the γ-ray

discriminator 32.

(5)   This embodiment can acquire precise

information on the position that γ-rays reach in the

25   radius direction of the through hole section 41 without

reducing the number of signal transmission substances

in the radiation detectors, and therefore using the

precise information on the position that γ-rays reach

improves the accuracy of tomography, eliminates the need for the reflector for the radiation detectors and can thereby prevent the reduction of the signal transmission substances, improve the energy resolution and suppress influences of scattered rays on the reconstruction of the tomography. As a result, this embodiment can improve the accuracy of tomography, that is, diagnostic accuracy of a PET inspection.

(6)   This embodiment uses semiconductor radiation detectors as the radiation detectors 5, and can thereby arrange a plurality of radiation detectors 5 in the radius direction of the through hole section 41 and such an arrangement of the plurality of radiation detectors 5 does not increase the size of the image pickup apparatus 2.

(7)   Using semiconductor radiation detectors for the radiation detectors 5, this embodiment eliminates the need for any photoelectron multiplier which is required for radiation detectors using a scintillator and can thereby reduce the size of the image pickup apparatus 2.

(8)   Arranging the radiation detectors 5 which are semiconductor radiation detectors on the support substrate, this embodiment allows the radiation detectors 5 to be arranged densely. Allowing especially radiation detectors 5 with small detector widths to be arranged densely in the circumferential direction of the through hole section 41, this

embodiment can realize high resolution (small image voxel size) of the tomographic image.

(9)   Adopting the configuration of arranging the radiation detectors 5 on the support substrate 6, this

5   embodiment allows the radiation detectors 5 to be arranged densely. This allows an arrangement of a plurality of radiation detectors 5 in the radius direction of the through hole section 41 in particular and realizes high detection efficiency. Moreover,

10   since the radiation detectors 5 in the radius direction can detect γ-rays independently, the resolution in the radius direction improves. In a 3D (three-dimensional) PET inspection in particular, there may be a case where γ-rays enter the radiation detectors 5 diagonally, but

15   with an improvement of the resolution in the radius direction, it is possible to grasp the direction of incidence of γ-rays accurately. This can improve the quality of PET images obtained.

(10)   According to this embodiment, wires connected

20   to the radiation detectors 5 are arranged in the support substrate 6, and therefore it is possible to shorten the distance between the radiation detectors 5 in the circumferential direction and axial direction of the through hole section 41. The shortening of the

25   distance between the radiation detectors 5 reduces omissions in γ-ray detection and substantially increases the γ-ray detection efficiency. The substantial increase of γ-ray detection efficiency can shorten a PET

inspection time.

(11)   Using the radiation detectors 5 which have detected γ-rays as the radiation detectors 5 to detect X-rays, the radiological imaging apparatus 1 need not

5   provide the radiation detectors 5 to detect X-rays and the radiation detectors 5 to detect γ-rays separately, and can thereby simplify the configuration and reduce the size of the apparatus. The radiation detectors 5 output both X-ray detection signals and γ-ray detection

10 signals.

(12)   In this embodiment, the X-ray detection section is positioned in an area formed between one end of the γ-ray detection section and the other end of the γ-ray detection section in the longitudinal direction of

15 the bed 20, and therefore even when the examinee 42 moves during an inspection independently of the movement of the bed 20, it is possible to combine the information of the first tomographic image (PET image) created based on the first information obtained from a

20 γ-ray detection signal outputted from the γ-ray detection section and the information of the second tomographic image (X-ray computer tomographic image) obtained from the X-ray detection signal outputted from the X-ray detection section and improve the accuracy of

25 the created tomographic image of the examinee 42. Using the tomographic image can improve the accuracy of diagnosis of the examinee. More specifically, the position and size of the affected area of cancer can be

recognized accurately. It is especially possible to diagnose cancer of lymph gland which is a small organ.

(13)   As described above, according to this embodiment, the radiation detection apparatus 43 consists of a plurality of radiation detectors 5 which output both X-ray detection signals and γ-ray detection signals (detection of X-rays to obtain X-ray detection signals is performed using the radiation detectors 5 to detect γ-rays to obtain γ-ray detection signals) and therefore the radiation detection apparatus 43 is provided with both functions of the γ-ray detection section and X-ray detection section. The radiation detection apparatus 43 can be said to arrange the γ-ray detection section and X-ray detection section coaxially. For this reason, this embodiment can reconstruct the first tomographic image at the position of an affected area (where PET radiopharmaceutical is concentrated) including internal organs and bones of the examinee 42 and reconstruct the second tomographic image including the image of the affected area of the examinee 42 using the γ-ray detection signals which are other output signals of the radiation detectors 5. The data of the first tomographic image and data of the second tomographic image are reconstructed based on the output signals of the radiation detectors 5 which detect both transmitted X-rays and γ-rays, and therefore it is possible to combine the data of the first tomographic image and the data of the second

tomographic image aligned with each other with high
accuracy. In this way, it is possible to easily obtain
an accurate tomographic image (combined tomographic
image) including images of the affected area, internal

5 organs and bones, etc. According to this combined
tomographic image, it is possible to accurately
identify the position of the affected area from the
relationship with the internal organs or bones. For
example, by aligning the data of the first tomographic

10 image and the second tomographic image based on the
central axis of the detector support member 8 (or
through hole section 41) of the image pickup apparatus
2, it is possible to easily obtain image data combining
both tomographic images.

15 (14)   In this embodiment, the X-ray detection
section detects X-rays 57 which have been radiated from
the X-ray source 17 and pass through the affected area
56 of the examinee 42 and the γ-ray detection section
detects γ-rays emitted caused by radiopharmaceutical

20 from the area (affected area) in the body of the
examinee 42 through which X-rays pass at the position
of the examinee irradiated with the X-rays, and
therefore it is possible to carry out an X-ray CT
inspection and PET inspection at the same position

25 without moving the examinee 42 with the bed 20. During
both inspections, the X-ray detection section outputs a
detection signal of X-rays that have passed through the
affected area 56 of the examinee 42 and the γ-ray

detection section outputs a detection signal of γ-rays emitted from the affected area 56. The first tomographic image data at the position of the affected area 56 obtained based on the X-ray detection signal is

5   combined with the second tomographic image data at the position of the affected area obtained based on the γ-ray detection signal, and therefore even if the examinee 42 can no longer keep the same posture moves on the bed 20 during an inspection, it is possible to

10   combine those tomographic image data pieces accurately. That is, this embodiment can obtain combined tomographic image data with a high degree of accuracy. Thus, using the combined tomographic image data (combined tomographic image) at the position of the

15   affected area 56 displayed on the display device 38 can improve the accuracy of diagnosis of the affected area 56. Even when the affected area is located especially in a place where organs intercross, the combined tomographic image data obtained in this embodiment

20   makes it possible to keep track of the position of the affected area appropriately and improve the accuracy of diagnosis of the affected area.

(15)   According to this embodiment, using the X-ray source axial transport apparatus (e.g., axial transport

25   arm 16), the X-ray source 17 can be transported in the axial direction of the radiation detector 5 during radiological inspection, and therefore an X-ray CT inspection can be carried out on the range of the

inspection target while carrying out a PET inspection on the range of the inspection target without transporting the examinee 42 in the axial direction of the radiation detection apparatus 43. When an X-ray CT

5   inspection is carried out on the range of the inspection target while moving the examinee 42 by transporting the bed 20 without moving the X-ray source 17 in the axial direction, the position of the area where PET radiopharmaceutical is concentrated also

10 moves in the axial direction. This means that the position at which a γ-ray pair is generated is moved in the axial direction, resulting in increased noise when PET image data is created, failing to obtain accurate PET image data. In this embodiment, the position at

15 which a γ-ray pair is generated is not moved in the axial direction, resulting in accurate PET image data and improved accuracy of combined tomographic image data.

(16)   According to this embodiment, it is possible

20 to detect a plurality of γ-ray pairs emitted from the examinee 42 using the radiation detectors 5 included in the radiation detection apparatus 43 and also detect X-rays which have been radiated from the X-ray source 17 moving in the circumferential direction and passed

25 through the examinee 42. In this way, while the conventional art requires an image pickup apparatus for detecting X-rays and another image pickup apparatus for detecting γ-rays, this embodiment only requires a single

image pickup apparatus to detect X-rays and γ-rays and simplifies the configuration of the radiological imaging apparatus capable of executing both X-ray CT inspection and PET inspection.

5   (17)   This embodiment allows the common radiation detectors 5 to obtain X-ray detection signals necessary for creating a first tomographic image and γ-ray detection signals necessary for creating a second tomographic image, making it possible to drastically

10   shorten the time required to inspect the examinee 42 (inspection time).   In other words, it is possible to obtain X-ray detection signals necessary for creating a first tomographic image and γ-ray detection signals necessary for creating a second tomographic image in a

15   short inspection time.   This embodiment eliminates the need in the prior art for transporting the examinee 42 from one image pickup apparatus for detecting transmitting X-rays to another image pickup apparatus for detecting γ-rays, and therefore further contributes

20   to a reduction of the inspection time of the examinee 42.

(18)   This embodiment rotates the X-ray source 17 and moves the radiation detection apparatus 43 in neither the circumferential direction nor axial

25   direction of the through hole section 41, can thereby reduce the capacity of the motor for rotating the X-ray source 17 compared to the motor necessary to move the radiation detection apparatus 43.   Power consumption

necessary for driving the motor of the latter can also be reduced compared to power consumption of the motor of the former.

(19)   Since γ-ray detection signals are inputted to
5   the X-ray signal processing apparatus 33, that is, the first signal processing apparatus is reduced drastically, it is possible to obtain an accurate first tomographic image data. Thus, using image data obtained by combining the first tomographic image data
10   and second tomographic image data allows the position of the affected area to be known precisely.

(20)   In this embodiment, the X-ray source 17 rotates inside the radiation detection apparatus 43, and therefore the inner diameter of the detector
15   support member 8 increases and the number of radiation detectors 5 that can be provided in the circumferential direction inside the detector support member 8 can be increased. An increase in the number of radiation detectors 5 in the circumferential direction results in
20   an improvement in sensitivity and resolution and improves the resolution of the tomographic image on the cross section of the examinee 42.

(21)   Since the axial transport arm 16 and X-ray source 17 are located inside the radiation detection
25   apparatus 43, they intercept γ-rays emitted from the examinee 42 during an X-ray CT inspection and the radiation detectors 5 located right behind them cannot detect the γ-rays, which may cause detection data

necessary for creation of a PET image to be lost.
However, since the X-ray source drive apparatus 15
rotates the X-ray source 17 and the axial transport arm
16 in the circumferential direction as described above,

5   the loss of data substantially causes no problem.
Especially, the rotational speed of the X-ray source 17
and the axial transport arm 16 is approximately 1 sec/1
slice, which is short enough when compared to a time
required for a PET inspection which takes approximately

10 a few minutes at the shortest. The loss of the data is
therefore substantially no problem. Furthermore, when
a PET inspection is carried out without any X-ray CT
inspection, the X-ray source 17 is housed in the X-ray
source drive apparatus 15 and therefore the X-ray

15 source 17 and the axial transport arm 16 do not
constitute obstacles to detection of γ-rays.

        Moreover, the inspection time necessary to
obtain an X-ray detection signal necessary to create an
X-ray CT image is shorter than the inspection time

20 necessary to obtain a γ-ray image pickup signal
necessary to create a PET image. Thus, during an
inspection time to obtain a γ-ray detection signal, X-
rays are always irradiated from the X-ray source 17
onto the examinee 42 to obtain an X-ray detection

25 signal and in this way even when the examinee 42 moves
during an inspection, it is also possible to correct a
data shift of the PET image caused by the movement of
the examinee 42 from successive X-ray CT images

obtained based on the X-ray detection signal.

The wires connected to the radiation detectors 5 are arranged inside the support substrate 6, but it is also possible to form a through hole in 5 the support substrate 6, pass the wires from the side of the support substrate 6 on which the radiation detectors 5 are arranged through the through hole, pull the wires out of the opposite side and arrange the wires on the surface of the support substrate 6 on the 10 side on which no radiation detectors 5 are arranged. In that case, it is also possible to form grooves on the surface of the side of the support substrate 6 on which no radiation detectors 5 are arranged and set the wires in the grooves. Furthermore, it is also possible 15 to use a multilayer wire board as the support substrate and set wires inside the multilayer wiring board. Furthermore, use of the multilayer wiring board allows the radiation detectors 5 to be arranged on both sides of the multilayer wiring board.

20 (Embodiment 2)

A radiological imaging apparatus according to Embodiment 2 which is another preferred embodiment of the present invention will be explained below. The radiological imaging apparatus in this embodiment only 25 differs from the configuration of the radiological imaging apparatus 1 of Embodiment 1 in the configuration of the detector unit 4. A detector unit 4A used in this embodiment having a configuration

different from that of the detector unit 4 used in

Embodiment 1 will be explained with reference to Fig.

10 and Fig. 11.

The detector unit 4A consists of a plurality

5   (e.g., 9) of radiation detectors 5D arranged in 3 rows

and 3 columns on one side of a support substrate 6.

Each radiation detector 5D is the same semiconductor

radiation detector as the radiation detector 5 and

three layers of the radiation detectors 5D are arranged

10 in the radius direction of a detector support member 8.

Of one column in the radius direction of the detector

support member 8, a radiation detector $5A_1$, radiation

detector $5B_1$ and radiation detector $5C_1$ are arranged on

the first layer, second layer and third layer

15 respectively.

The radiation detector $5A_1$ consists of five

detection elements, that is, detection elements 74A,

74B, 74C, 74D and 74E. The detection elements 74A,

74B, 74C, 74D and 74E are arranged in that order from

20 the inner side in the radius direction of the detector

support member 8. The detection element 74A is

arranged on the innermost side, while the detection

element 74E is arranged on the outermost side. A

cathode electrode 77A is provided on the inner side of

25 the detection element 74A. The detection element 74A

and detection element 74B are adjacent to each other

with an anode electrode 78A sandwiched between the

outer surface of the detection element 74A and the

inner surface of the detection element 74B. The
detection element 74B and detection element 74C are
adjacent to each other with a cathode electrode 77B
sandwiched between the outer surface of the detection

5   element 74B and the inner surface of the detection
element 74C. The detection element 74C and detection
element 74D are adjacent to each other with an anode
electrode 78B sandwiched between the outer surface of
the detection element 74C and the inner surface of the

10   detection element 74D. The detection element 74D and
detection element 74E are adjacent to each other with a
cathode electrode 77C sandwiched between the outer
surface of the detection element 74D and the inner
surface of the detection element 74E. An anode

15   electrode 78C is provided on the outer surface of the
detection element 74E.

The radiation detector $5B_1$ consists of five
detection elements, that is, detection elements 75A,
75B, 75C, 75D and 75E. The detection elements 75A,

20   75B, 75C, 75D and 75E are arranged in that order from
the inner side in the radius direction of the detector
support member 8. The detection element 75A is
arranged on the innermost side, while the detection
element 75E is arranged on the outermost side. A

25   cathode electrode 79A is provided on the inner side of
the detection element 75A. The detection element 75A
and detection element 75B are adjacent to each other
with an anode electrode 80A sandwiched between the

outer surface of the detection element 75A and the inner surface of the detection element 75B. The detection element 75B and detection element 75C are adjacent to each other with a cathode electrode 79B

5 sandwiched between the outer surface of the detection element 75B and the inner surface of the detection element 75C. The detection element 75C and detection element 75D are adjacent to each other with an anode electrode 80B sandwiched between the outer surface of

10 the detection element 75C and the inner surface of the detection element 75D. The detection element 75D and detection element 75E are adjacent to each other with a cathode electrode 79C sandwiched between the outer surface of the detection element 75D and the inner

15 surface of the detection element 75E. An anode electrode 80C is provided on the outer surface of the detection element 75E.

The radiation detector $5C_1$ consists of five detection elements, that is, detection elements 76A,

20 76B, 76C, 76D and 76E. The detection elements 76A, 76B, 76C, 76D and 76E are arranged in that order from the inner side in the radius direction of the detector support member 8. The detection element 76A is arranged on the innermost side, while the detection

25 element 76E is arranged on the outermost side. A cathode electrode 81A is provided on the inner side of the detection element 76A. The detection element 76A and detection element 76B are adjacent to each other

with an anode electrode 82A sandwiched between the

outer surface of the detection element 76A and the

inner surface of the detection element 76B. The

detection element 76B and detection element 76C are

5   adjacent to each other with a cathode electrode 81B

sandwiched between the outer surface of the detection

element 76B and the inner surface of the detection

element 76C. The detection element 76C and detection

element 76D are adjacent to each other with an anode

10   electrode 82B sandwiched between the outer surface of

the detection element 76C and the inner surface of the

detection element 76D. The detection element 76D and

detection element 76E are adjacent to each other with a

cathode electrode 81C sandwiched between the outer

15   surface of the detection element 76D and the inner

surface of the detection element 76E. An anode

electrode 82C is provided on the outer surface of the

detection element 76E.

A grounding wire 45 is connected to the

20   cathode electrodes 77A, 77B, 77C, 79A, 79B, 79C, 81A,

81B and 81C. A wire 74 is connected to the anode

electrodes 78A, 78B and 78C. A wire 75 is connected to

the anode electrodes 80A, 80B and 80C. A wire 76 is

connected to the anode electrodes 82A, 82B and 82C.

25   The grounding wire 45 is connected to a connector

terminal 7D of the connector section 7. The wire 74 is

connected to a connector terminal 7A of the connector

section 7. The wire 75 is connected to a connector

terminal 7B of the connector section 7. The wire 76 is
connected to a connector terminal 7C of the connector
section 7. The radiation detectors 5D included in
other two columns are likewise connected to other
5   connector terminals provided for the connector section
7. All of the grounding wires 45, 74, 75 and 76 are
set in the support substrate 6. Many detector units 4A
are attached to a detector support section 23 and held
by fitting connector terminals such as the connector
10   terminal 7A provided respectively into the connector
section 11 provided for the detector support section
23. As in the case of the detector units 4, many
detector units 4A are arranged surrounding the through
hole section 41 in the circumferential direction and
15   axial direction of the through hole section 41.

      The radiation detectors $5A_1$, $5B_1$ and $5C_1$
comprise three or more detection elements having at
least two surfaces, that is, semiconductor elements and
arrange anode electrodes and cathode electrodes
20   alternately between different semiconductor elements.
More specific explains will be given using the
radiation detectors $5A_1$. The radiation detector $5A_1$
comprises anode electrodes and cathode electrodes
alternately between different detection elements, that
25   is, between the detection element 74A and detection
element 74B, between the detection element 74B and
detection element 74C, between the detection element
74C and detection element 74D and between the detection

element 74D and detection element 74E, for example, the
anode electrode 78A between the detection element 74A
and detection element 74B, or the cathode electrode 77B
between the detection element 74B and detection element

5   74C.

When the connector section 7 is engaged with
the connector section 11, the three radiation detectors
$5A_1$ on the first layer are connected to three signal
discriminators 27 in a signal discrimination unit 25

10 separately. Furthermore, the three radiation detectors
$5B_1$ on the second layer and the three radiation
detectors $5C_1$ on the third layer are connected to six γ-
ray discriminators 32 other than the signal
discriminators 27 provided in the signal discrimination

15 unit 25 separately.

The radiological imaging apparatus of this
embodiment incorporating the radiation detector units
4A can achieve the effects (1) to (21) produced by the
radiological imaging apparatus 1 of Embodiment 1.

20 Furthermore, this embodiment can achieve effects (22)
and (23) shown below.

(22)   According to this embodiment, the radiation
detector 5D has a multilayered structure of a plurality
of detection elements, which reduces the thickness of

25 the respective detection elements between the anode
electrode and cathode electrode and suppresses
reductions of detection signals due to re-coupling
between electrons and holes. This improves energy

resolution. Furthermore, the time until a detection signal is outputted is shortened and so the time resolution also improves. With the improved energy resolution, it is possible to set a high energy

5 threshold and thereby remove more γ-rays whose energy has decreased due to scattering. Moreover, with the improved time resolution, it is possible to reduce the time window and thereby reduce γ-rays which are detected accidentally within the time window. That is, it is

10 possible to suppress scattering phenomena and accidental phenomena which constitute noise components to a low level and thereby improve the image quality of a PET image.

(23)   According to this embodiment, the radiation

15 detector 5D has a multilayered structure of a plurality of detection elements, and therefore the thickness of detection elements between the anode electrode and cathode electrode is reduced and a bias voltage to be applied can be reduced. With a reduced bias voltage,

20 it is possible to reduce a withstand voltage of parts around various wires. Furthermore, the size of the power supply itself can be reduced.

(Embodiment 3)

A radiological imaging apparatus according to

25 Embodiment 3 which is another preferred embodiment of the present invention will be explained below. The radiological imaging apparatus in this embodiment only differs from the configuration of the radiological

imaging apparatus 1 of Embodiment 1 in the configuration of the detector unit. A detector unit 4B used in this embodiment having a configuration different from that of the detector unit 4 used in

5   Embodiment 1 will be explained with reference to Fig. 12 and Fig. 13.

The detector unit 4B consists of a plurality (e.g., 9) of radiation detectors 5E arranged in 3 rows and 3 columns on one side of a support substrate 6.

10 Each radiation detector 5E is the same semiconductor radiation detector as the radiation detector 5 and three layers of the radiation detectors 5E are arranged in the radius direction of a detector support member 8. In the radius direction of the detector support member

15 8, radiation detectors 5Aa, 5Ab and 5Ac, radiation detectors 5Ba, 5Bb and 5Bc, and radiation detectors 5Ca, 5Cb and 5Cc are arranged on the first layer, second layer and third layer respectively. The radiation detectors 5Aa, 5Ab and 5Ac arranged in the

20 circumferential direction of the detector support member 8 are multilayered in the circumferential direction thereof. The radiation detectors 5Ba, 5Bb and 5Bc on the second layer and radiation detectors 5Ca, 5Cb and 5Cc on the third layer are likewise

25 multilayered in the circumferential direction of the detector support member 8. This multilayered structure will be explained by taking the radiation detectors 5Aa, 5Ab and 5Ac as an example.

The radiation detector 5Aa comprises four detection elements, that is, detection elements 83A, 83B, 83C and 83D. The detection elements 83A, 83B, 83C and 83D are arranged in that order in the

5   circumferential direction of the detector support member 8. The radiation detector 5Ab comprises detection elements 84A, 84B, 84C and 84D. The detection elements 84A, 84B, 84C and 84D are arranged in that order in the circumferential direction of the

10   detector support member 8. The radiation detector 5Ac comprises detection elements 85A, 85B, 85C and 85D. The detection elements 85A, 85B, 85C and 85D are arranged in that order in the circumferential direction of the detector support member 8.

15   A cathode electrode 86A is provided on one side of the detection element 83A. The detection element 83A and detection element 83B are adjacent to each other with an anode electrode 87A sandwiched between the other side of the detection element 83A and

20   one side of the detection element 83B. Here, one side refers to one side of the detection element in the circumferential direction of the detector support member 8 and the other side refers to the remaining side of the detection element in the circumferential

25   direction of the detector support member 8. The detection element 83B and detection element 83C are adjacent to each other with a cathode 86B provided between the other side of the detection element 83B and

one side of the detection element 83C. The detection

element 83C and detection element 83D are adjacent to

each other with an anode 87B provided between the other

side of the detection element 83C and one side of the

5   detection element 83D.

The detection element 83D and detection

element 84A are adjacent to each other with a cathode

electrode 88A provided between the other side of the

detection element 83D and one side of the detection

10   element 84A. The detection element 84A and detection

element 84B are adjacent to each other with an anode

electrode 89A provided between the other side of the

detection element 84A and one side of the detection

element 84B. The detection element 84B and detection

15   element 84C are adjacent to each other with a cathode

electrode 88B provided between the other side of the

detection element 84B and one side of the detection

element 84C. The detection element 84C and detection

element 84D are adjacent to each other with an anode

20   electrode 89B provided between the other side of the

detection element 84C and one side of the detection

element 84D.

The detection element 84D and detection

element 85A are adjacent to each other with a cathode

25   electrode 90A provided between the other side of the

detection element 84D and one side of the detection

element 85A. The detection element 85A and detection

element 85B are adjacent to each other with an anode

electrode 91A provided between the other side of the detection element 85A and one side of the detection element 85B. The detection element 85B and detection element 85C are adjacent to each other with a cathode

5   electrode 90B provided between the other side of the detection element 85B and one side of the detection element 85C. The detection element 85C and detection element 85D are adjacent to each other with an anode electrode 91B provided between the other side of the

10 detection element 85C and one side of the detection element 85D. A cathode electrode 90C is provided on the other side of the detection element 85D.

   A grounding wire 92 is connected to the cathode electrodes 86A, 86B, 88A, 88B, 90A, 90B and

15 90C. A wire 93 is connected to the anode electrodes 87A and 87B. A wire 94 is connected to the anode electrodes 89A and 89B. A wire 95 is connected to the anode electrodes 91A and 91B. The grounding wire 92 is connected to a connector terminal 7D of a connector

20 section 7. The wire 93 is connected to a connector terminal 7A of the connector section 7. The wire 94 is connected to a connector terminal 7B of the connector section 7. The wire 95 is connected to a connector terminal 7C of the connector section 7. The radiation

25 detectors 5E on the second layer and third layer are likewise connected to other connector terminals provided on the connector section 7. All of the grounding wire 92 and wires 93, 94 and 95 are set

inside the support substrate 6. Many detector units 4B are attached to the detector support section 23 and held by fitting connector terminals such as the connector terminal 7A provided respectively into the

5   connector section 11 provided for the detector support section 23. As in the case of the detector units 4, the detector units 4B are arranged surrounding the through hole section 41 in the circumferential direction and axial direction of the through hole

10   section 41.

When the connector section 7 is engaged with the connector section 11, the radiation detectors 5Aa, 5Ab and 5Ac on the first layer are connected to three signal discriminators 27 in the signal discrimination

15   unit 25 separately. The radiation detectors 5Ba, 5Bb and 5Bc on the second layer and the radiation detectors 5Ca, 5Cb and 5Cc on the third layer are connected to six γ-ray discriminators 32 other than the signal discriminator 27 provided in the signal discrimination

20   unit 25 separately.

The radiological imaging apparatus incorporating the detector units 4B of this embodiment can achieve the effects (1) to (21) produced by the radiological imaging apparatus 1 of Embodiment 1 and

25   the effects (22) and (23) produced by the radiological imaging apparatus of Embodiment 2. Furthermore, this embodiment can achieve the following effect (24).

(24)   According to this embodiment, each radiation

detector 5E has a multilayered structure of a plurality
of detection elements, and therefore it is possible to
use both sides of the adjacent radiation detectors 5E
as cathode electrodes and allow those radiation

5   detectors 5E to share the cathode electrodes. This
allows the three radiation detectors 5E arranged in the
circumferential direction of the detector support
member 8 to be arranged close to one another. That is,
it is possible to completely eliminate spaces between

10   the radiation detectors 5E in the circumferential
direction and considerably reduce omissions in γ-ray
detection between the radiation detectors 5E in the
circumferential direction. This leads to a substantial
increase in γ-ray detection efficiency and shortening of

15   an inspection time as well.

Embodiments 1 to 3 have a configuration in
which detector units comprising radiation detectors
arranged in multilayers in the radius direction of the
detector support member 8 (through hole section 41 into

20   which the bed 20 is inserted) are applied to a
radiological imaging apparatus which can irradiate X-
rays from the X-ray source 17 onto an examinee and
detect γ-rays and X-rays. However, the detector units
can also be applied to a PET radiological imaging

25   apparatus which does not irradiate X-rays but only
detects γ-rays emitted from an examinee caused by
radiopharmaceutical which has passed through the
examinee. Furthermore, the detector units can also be

applied to a SPECT radiological imaging apparatus.

The present invention can improve the accuracy of images created and facilitate replacement of damaged radiation detectors.

5   It should be further understood by those skilled in the art that although the foregoing description has been made on embodiments of the invention, the invention is not limited thereto and various changes and modifications may be made without

10   departing from the spirit of the invention and the scope of the appended claims.